Patient-specific FE model of the leg under elastic compression

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1. ABSTRACT

Elastic compression (EC) is a medical treatment which relies on the use of socks to improve the venous return and, thereby, control or treat various vein-related diseases such as ulcers. The beneficial effects of EC have been known for centuries, but their mechanism of action is not totally understood. In order to validate and improve current treatments, it is necessary to determine how the pressure is transmitted from the sock to the vein walls.

To address this issue, a patient-specific 3D FE model of the leg under EC was created for a group of six subjects. CT-scans of each patient were segmented into three regions, namely the superficial soft tissues (mostly adipose and skin), the deep soft tissues (mostly muscles) and the bones, to create the geometry of each model. The local pressure applied by the sock on the skin was estimated using Laplace's law. The hard tissues, considered here as not deformable, were fixed in the model. Soft tissues, namely the adipose tissue and muscle, were defined as isotropic and hyper-elastic (Neo-Hookean strain energy function). The mechanical properties were identified in 3D by an inverse method to fit the FE models on the CT-scans of the legs under EC.

The main result is that the mean pressure applied by the EC onto the skin is similar to the pressure applied by the compressed tissues onto the wall of the three main deep veins. This suggests that the mean pressure applied can be used as an indicator of the efficiency of the EC. In a similar way, the maximal hydrostatic pressure can be used to estimate the comfort. Indeed, the results showed that this pressure is inversely proportional to the adipose tissues thickness.

2. INTRODUCTION

Compression therapy is a very effective modality for controlling and treating venous disorders of the lower leg such as varicose veins and ulcers. It has been correlated with both physiological effects on venous blood flow and clinical benefits. In practical applications however, it has been observed that the response of the internal tissues of the calf to external compression is highly variable, thereby impacting on the effectiveness of the treatment. Conflicting results have been reported for example regarding the beneficial effect of EC on venous hemodynamics and on the resulting efficacy of EC

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therapy to achieve the desired medical goals. This highlights the strong need to improve the current understanding of the mechanisms by which elastic compression (EC) stockings participate in the management of venous insufficiency.

Developments in that respect have focused on two main approaches, namely the evaluation of the hemodynamic response of the leg to EC [1, 2] and the assessment of the contact pressure performance of elastic stockings [3, 4] both experimentally and numerically. However, very few studies have addressed the essential question of the transmission of pressure through the soft tissues of the leg and the resulting stress distribution which is very likely to condition the performance of EC in achieving the medical goals [7, 7]. The aim of the present study is to model the pressure distribution in the calf resulting from external compression on a significant number of patients.

3. METHOD

3.1 Protocol

In this study, the medical images of the leg of six voluntaries were acquired. The protocol followed consisted of the following three steps. First, the volunteers signed an informed consent form, according to a protocol approved by the local institutional ethics committee. A medical examination was then carried out on all the volunteers to make sure that neither of them had venous return diseases. Their leg perimeters were measured to prescribe the right EC socks according to the manufacturer's recommendation. *Pro Recup* EC socks from the BVSport® society were used. Finally, the subjects did a CT-scan imaging of one of their leg with and without EC.

3.2 Image processing

Segmentation

The 3D CT-scan images were segmented into three regions (Fig. 1):

1. the superficial soft tissues, composed principally of adipose tissues, skin and some veins,
2. the deep soft tissues, composed principally of muscles, tendons and blood vessels,
3. the hard tissues, consisting of the two bones (the tibia and the fibula).

The segmentation was performed using the ImageJ software. A 3D visualisation of the segmentation for each subject is shown in Figure 1.

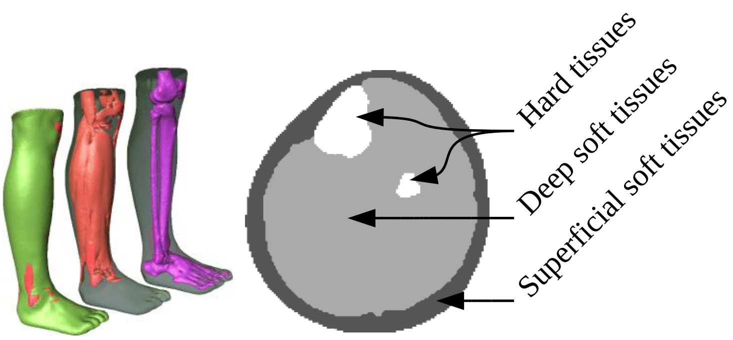


Fig. 1: Segmentation

Image warping

The images of the legs with and without EC needed to be in the same reference mark so that an inverse method could be used in the later steps. In order to identify the rigid transformation of the leg between the two images, a least-squares method was used to superimpose the hard tissues of both legs according to the formula:

$$[\hat{R},\hat{T}] = arg_{(R,T)}min\sum_{i}^{N}\|Q_i - RP_i - T\|$$

where R and T are the optimal rotation and translation respectively to superimpose the two images, N is the pixel number in the images, Q_i and P_i are the coordinate of the leg without and with EC, respectively. Only the two bones were used to perform image warping because it is assumed that only the bones are not deformed under EC.

3.3 Finite element model

Meshing

The segmented images were meshed with linear tetrahedral elements (approximately 500,000 elements and 60,000 nodes for each model).

Boundary conditions

The bones were supposed not deformable compared to the soft tissues, thus the hard tissues were fixed.

A pressure modelling the action of the EC sock was applied on the leg. This pressure, P, was computed with the Laplace's law:

$$P = Stiff\frac{\varepsilon}{Rc} \quad (1)$$

where *Stiff* is the stiffness of the sock textile (estimated from a previous study [7]), Rc the curvature radius of the leg with EC (to have the pressure in the final state) in the horizontal plane and ε is the strain of the sock in the horizontal plane. The latter was derived knowing leg and sock perimeters from the CT-scans.

Constitutive equation

Both soft tissues were assumed homogeneous, isotropic, quasi-incompressible and governed by a neo-hookean strain energy function, φ:

$$\varphi = c_{10}(\bar{I_1} - 3) + \frac{\kappa}{2}(J-1)^2 \quad (2)$$

where c_{10} and κ are the neo-hookean parameters driving the constitutive equation, $\bar{I_1} = Tr(\bar{\boldsymbol{F}}.\bar{\boldsymbol{F}}^t)$ is the first deviatoric strain invariant and $J = \det(\boldsymbol{F})$ is the volume ratio. The κ parameter was fixed at 1 kPa for both soft tissues and the parameter c_{10} was identified by an inverse method.

3.4 Identification

Inverse method

The inverse method is schematized in Fig. 2. First, a FE model is created from the initial experimental data (CT-scans of legs without EC). The result of the simulation is compared with the final experimental data (CT-scans of legs with EC) by a cost function. This cost function, detailed bellow, estimates the difference between the target and the simulated leg. Then, an optimisation algorithm computes the new constitutive parameters and the loop starts again. When the cost function's minimum is found, the constitutive parameters are considered to be identified.

The used optimisation algorithm is the Nelder Mead algorithm (implemented in Matlab® in the *fminsearch* function).

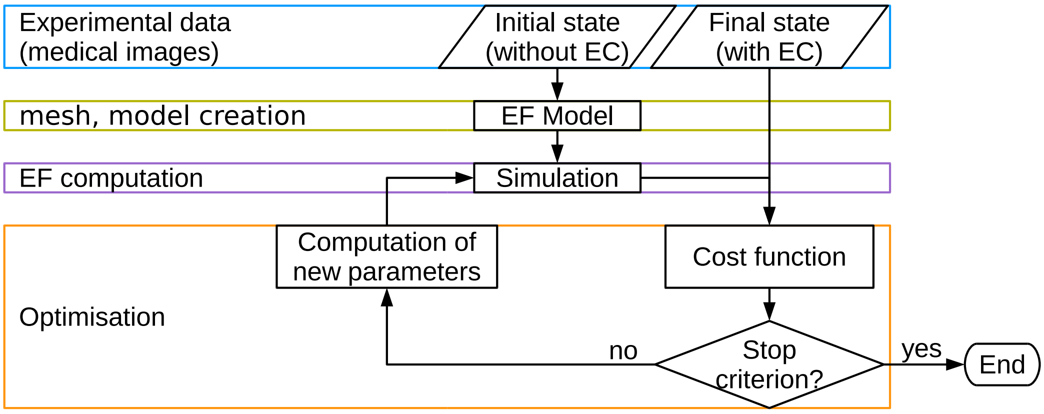
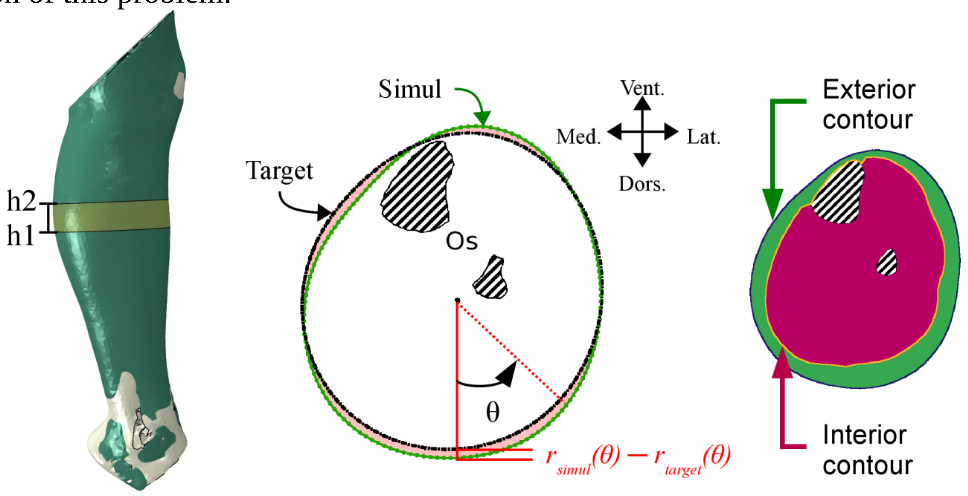


Fig. 2: Inverse Method

Cost function
The cost function used depends on the shape of the leg contours:

$$C = \sum_{z=h1}^{h2} \sum_{\theta} \left(\left[\frac{r_{simul}(z,\theta) - r_{target}(z,\theta)}{r_{target}(z,\theta)} \right]_{ext.}^{2} - \left[\frac{r_{simul}(z,\theta) - r_{target}(z,\theta)}{r_{target}(z,\theta)} \right]_{int.}^{2} \right) \qquad 3$$

where $r_{simul}(z,\theta)$ and $r_{target}(z,\theta)$ are, respectively, the simulation and the target radius at the θ angle of the contours (Fig. 3b), for the outer (exterior) and inner (interior) contours of the superficial soft tissues (Fig. 3c) at the height z of the leg; h_1 and h_2 are the boundary heights for the identification (Fig. 3a).
This cost function was chosen following a precedent study. It allows having a unique solution of this problem.



(a) (b) (c)

Fig. 3: Cost function. Minimisation of the difference of the target and simulation radius for all the contour of the leg (b) (minimisation of the red area on the scheme), for five slices between the heights $h_1 - h_2$ (a), and for the interior and exterior contours (c).

4. RESULTS

4.1 Identification

The c_{10} value identified for the deep soft tissue was 3.25±0.93 kPa. This result is in the same order of magnitude than the value obtained by [1] for an *in vivo* indentation of a pig muscle (4.25 kPa). The value for the superficial soft tissue was 8.17±7.22 kPa, which is closed to the value used by [2] for the buttock fat (11.7 kPa).

4.2 Pressure

The aim of this study is to characterize the pressure transferred from the compression garment to the soft tissues. Thus, the hydrostatic pressure was analysed. This quantity was chosen for the analysis because it does not depend on the coordinate system and it could be used to predict local fluid flows. It is assumed to equal the pressure applied by the tissue onto the wall of a vein at the same location. Results are displayed in Figure 4 for two subjects. Whereas the sock size is adapted for each subject, the models show large inter-individual variability of the pressure field.

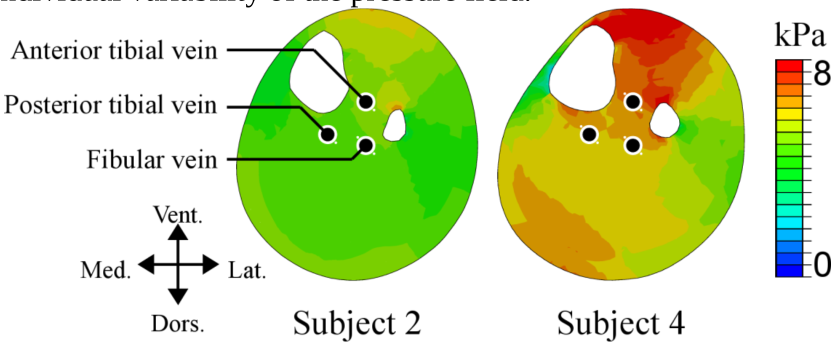


Figure 4: Pressure field for two subjects in the leg cross section and the deep vein locations.

The pressure at the deep vein locations (Figure 4) is similar to the applied pressure (averaging in the circumference) for almost all the subjects (Figure 5). This result indicates that the deep veins are subject to the same pressure as the one applied externally on the skin. Thereby, the mean applied pressure could be an indicator of the EC efficiency. Indeed, the leg venous return depends on the flux in the deep veins because the later are the bigger veins of the leg. Furthermore, the mean applied pressure could be estimated by the leg perimeter with the Laplace law (using the mean radius of the leg).

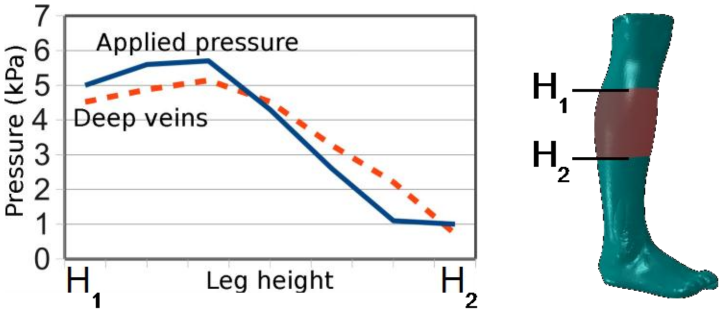


Figure 5: Mean (in the circumference) applied pressure and mean hydrostatic pressure in the main deep veins locations

Also, the models show that there are significant local variations of the maximal hydrostatic pressure in the soft tissue between the subjects, and this seems to depend linearly on the adipose tissue thickness. Considering that the maximal hydrostatic pressure could be related to comfort [9], this result suggests a simple way to estimate the EC comfort of patients by measuring the adipose tissue thickness.

3 CONCLUSIONS

In the present work, a 3D FE model of the leg was developed to analyse the

biomechanical response of the leg to external compression in a group of six subjects. The non-linear properties of the soft tissues were identified using an inverse method. Using the model, it has been shown that the spatial variations in pressure are significant across the leg. It has also been shown that the mean pressure applied by the EC onto the skin is similar to the pressure applied by the compressed tissues onto the wall of the three main deep veins. Nevertheless, it has been observed that the maximal hydrostatic pressure is directly proportional to the adipose tissue thickness. This indicates that adipose tissue thickness is an indicator of comfort.
Yet, the relationship between the hydrostatic pressures measured in the vicinity of the veins using the developed model and the impact on vein-related diseases needs to be further investigated to improve our current understanding of the mechanisms of action of EC. To address the latter point, a more realistic model will be implemented in 2D to analyse the local response of the vein wall. Refinement of the muscle models are also under progress

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